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Two-Lasers Assisted Ablation: A Method for Enhancing Conventional Laser Ablation of Materials

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A new method for enhanced ablation by pulsed laser radiation has been demonstrated. The method utilizes an "impulse" ablation laser in conjunction with cyclical heating of the tissue by an auxiliary source: the "primer." In this study we use an auxiliary laser as the "primer" heat source, which sets up a thermal and stress field modulation in the target material. The cyclical stress associated with the thermal modulation reduces material strength thus enhancing ablation (larger mass removed per pulse) by the "impulse" laser. Ablation rate enhancement of more than twice the single-laser value is demonstrated.

Key words: **enhanced material removal, thermal modulation, tissue interaction, two-laser ablation**

INTRODUCTION

Laser interaction with material and biological tissue has been a subject of ongoing research for the past 35 years. One of the most important problems in ablation theory is that of maximizing ablation rates (AR), while simultaneously minimizing collateral damage, enhancing selectivity, and improving precision of the interaction process. This problem is shared by both the biomedical application research as well as the material processing field. Traditional means for achieving these goals include either the enhancement of absorption through wavelength selection or the application of auxiliary absorbers to the target surface. Other options include manipulations of additional laser parameters such as energy, spot size, pulse duration, pulse repetition rates, and total delivery time.

Two-beam ablation has been reported previously. Fox [1] investigated a method for increasing the penetration efficiency of CO_2 CW laser beam by combining it with Q-switched Nd: glass which decreased ablation rates of a steel targets by more than a factor of 2. Hole was also found to be substantially of resolidified fragments which are commonly associated with CW penetration. Fox attributed the enhanced ablation and cleaner surfaces generated by his method to the cleaning of residual depositions by to the pulse laser action. Similarly, Robin and Nordin [2] concluded that improved CW laser penetration of solids can be affected using a superimposed pulsed laser operation. They suggested that the effect was due to the removal of melt due to a blow off generated by the short superimposed laser pulses.

Kim et al. [3] have investigated the effect of chopping on laser penetration of metal targets. By amplitude modulating a free running Nd:YAG laser pulse the depth of the crater was increased and threshold energy was decreased, for optimal operation a 8–12 kHz operation. They concluded that the improved in ablation rates was due to reduced plasma screening and possibly due to acoustics resonance effects similar to the disintegration of water due to acoustic vibration.

Finally, Izatt et al. [4] investigated ablation process in bone using two pulses separated by veritable time delays. They showed that for two sub-

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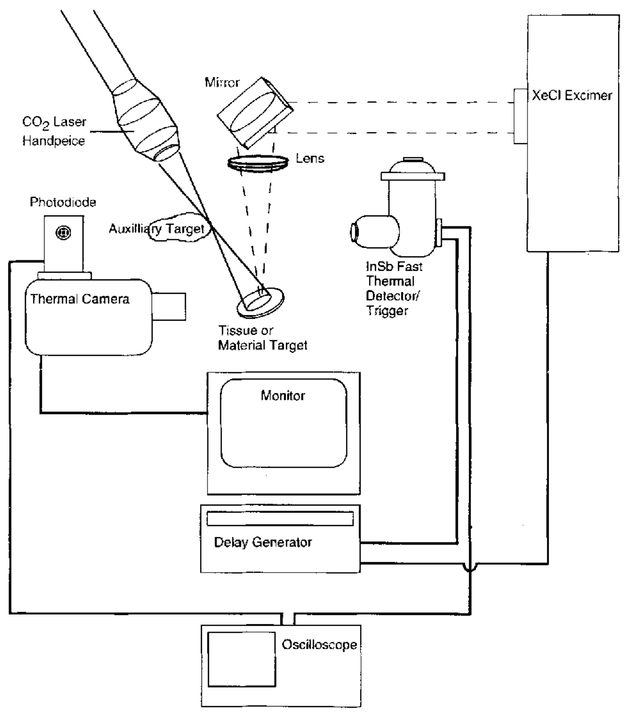

Fig. 1. CHAA set up

ablation threshold, 7.5 ns 355 nm pulses crater depth was not affected by time delays longer than 100 ns. Crater depth decreased monotonically to zero following time delays of 10 ns. In other experiments where the second pulse was then replaced by one from a 532 nm laser source, similar crater qualities were observed.

In this report we describe a novel procedure for effecting enhanced material ablation through the simultaneous, synchronized application of two lasers. The two-laser system serves as an example for what we believe is a more general technique which we may call cyclic-heating assisted ablation (CHAA) where one of the two lasers could be replaced by an alternative localized modulated heat source.

In its current configuration (Fig. 1), our method combines the synchronous operation of two separate laser sources: a "primer" (P) laser and an "impulse" (I) laser. The "primer" laser operates at a low fluence level (below the threshold for damage). Its fluence level, however, must be sufficiently high to induce cyclic temperatures and modulated stress in the material. The second laser, the "impulse" laser, is designed to provide sufficient energy to compromise tissue integrity and achieve subsequent ablative removal of the material. The impulse laser energy may be greater or smaller than the ablation threshold.

The primer laser selection depends on the thermal relaxation time (τ_r) which is given by

$$\tau_r = (1/\alpha)^2/4\kappa = \delta^2/4\kappa$$

τ_r, in turn, depends on the optical penetration depth (δ) and the material thermal diffusivity (κ). To demonstrate the utility of the procedure, we calculated τ_r for high-water-content media. For CO_2 lasers, emitting the most commonly used wavelength (10.6 μm) in applications in surgery, the τ_r value is 114 μs. Again, note that our objective is to generate sub ablative thermally-induced stress in the medium. The P-laser was thus selected to operate below damage threshold. By allowing the target sufficiently long time to remove the heat before the next P-laser pulse arrives, we minimize steady-state temperature increases. Indeed, thermal camera measurements (Interferometric Model 600, Bedford, MA) confirmed temperature increases of only 10°C (for dentin) to 25°C (for inorganic material), well below permanent damage or denaturation.

As our primer laser (see experiment set up configuration in Fig. 1), we selected a Superpulse CO_2 laser (Xanar, Inc. XA-50, Colorado Springs, CO) with the micropulse duration of 150 μs and a pulse to pulse separation of about 3 ms. This ensured complete thermal relaxation between pulses. The time duration of the entire macropulse can be varied. For dentin ablation we selected approximately 200 ms long macropulse which contain 67 micropulses (Fig. 2). Impulse-laser pulse was delivered 100 ms after the start of the macropulse. Enhance glass ablation was demonstrated with a shorter macropulse of 10 ms. Each micropulse energy was 35 mJ. Off time between micropulses was 2.85 ms. Only a single macropulse was used for priming in each ablation experiment. The beam profile was Gaussian and spot size was 1.1 cm^2 (1/e = 5.9 mm). We note, however, that the relationship of the entire macropulse duration to the exact timing of the "impulse" impact is not clear yet and is currently the subject of further experiments on which we hope to report soon.

We used a fast InSb detector (Cincinnati Electronics Corp., SDD-1963-S1-05M, Cincinnati, OH) to monitor the thermal cycles in an auxiliary target which used a portion of the Primer laser beam. The InSb detector also triggered a time delay generator (TDG). The TDG (Stanford Research, Model GEN-535, Palo Alto, CA) was used to trigger the Impulse laser which effected material expulsion. The auxiliary target was used be-

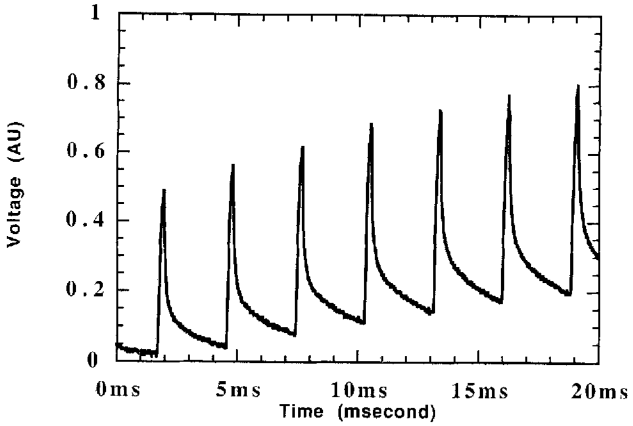

Fig. 2. "Primer" laser macropulse temporal structure (total macropulse duration is 200 ms; Individual micropulses duration is 150 μs).

cause the thermal effect generated by the Impulse laser was also sufficient to trigger the InSb detector. Monitoring the onset of the Primer pulse on a separated target avoided secondary triggering. The I-laser was chosen so that its short pulse duration indeed approximated an impulse (delta function) compared to the target thermal relaxation time and heating cycles. We first selected a 15 ns XeCl excimer laser.

The irradiation parameters for the impulse laser in the case of dentin ablation were as follows: A 15 ns XeCl excimer laser (Lumonics HyperEX-400, Kanata, Canada) with a flat top rectangular beam (1×3 mm^2) was used. A single 15 ns pulse was used synchronously to the time delay generator and the primer laser. Energy was maintained at 13.4 mJ/pulse 450 mJ/Cm^2, which is not much above the single pulse fluence ablation threshold for this laser [5]. Very clear ablation enhancement was demonstrated in this case (Fig. 3a). For the P-laser average power of 12 W the temperatures observed by our thermal camera reached only 35°C, well below the threshold for physical modification of the tissue. When the I-laser operated alone, its ablation rate (AR) was 1.09 +/− 0.14 μm/pulse (n = 5). When the CO_2 was activated AR jumped to a remarkable 2.43 +/− 0.24 μm/pulse (again n = 5). Note that the same tested sample was used in an alternating manner for single/dual laser operation to ensure that variations due to individual tissue characteristics were minimized).

The effect of CHAA was then demonstrated again in Dehydrated Dentin samples (Fig. 3b). I-laser parameters were maintained as above, but P-laser energy was increased by a factor of six.

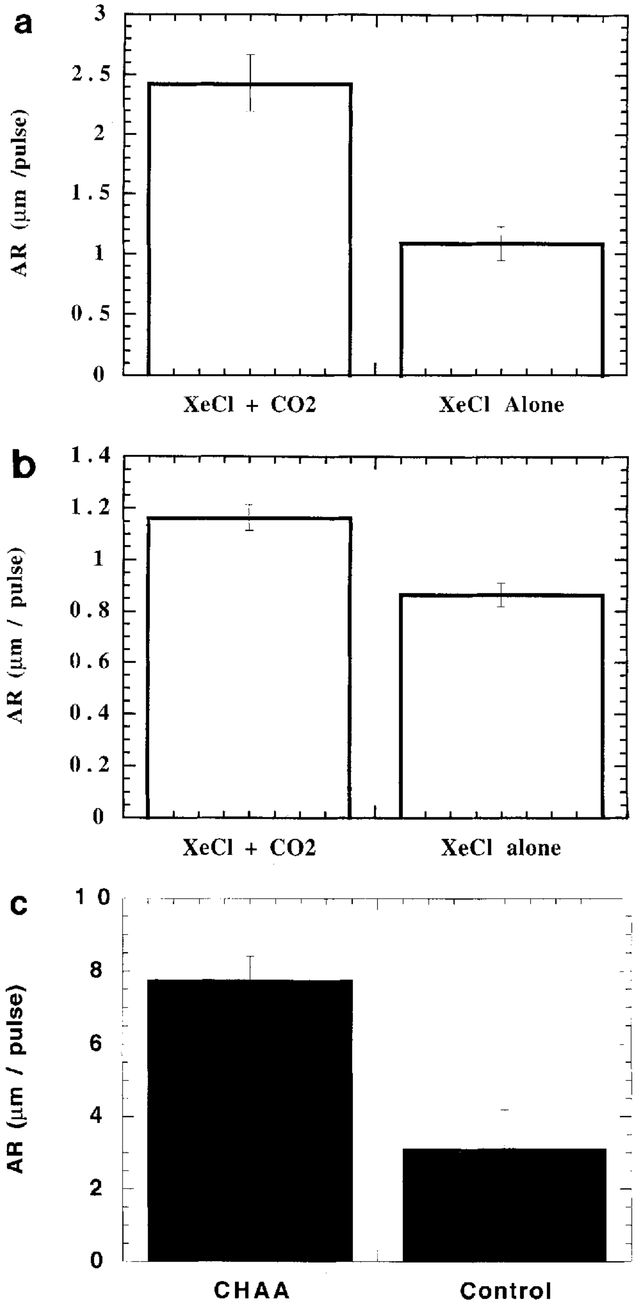

Fig. 3. **a:** Ablation rate per pulse in fresh dentin with and without CHAA. Fig. 3. **b:** Ablation rate per pulse in dehydrated dentin with and without CHAA. Fig. 3. **c:** Ablation rate per pulse in glass with and without CHAA.

(Recall, though, that our calculations of thermal relaxation time were based on the water-saturated tissue as an example.) AR enhancement was again demonstrated: 0.86 +/− 0.05 μm/pulse I-laser alone, compared to 1.16 +/− 0.05 μm/pulse for CHAA.

The clearest demonstration of CHAA with the larger sample size, (n = 25) was illustrated on a non-organic material by the ablation of glass

plates (Fig. 3c). To demonstrate that ablation enhancement could be achieved for a wide range of I-laser characteristics, we used in this case a Mode-locked Nd:YAG 1.06 μm (Antares 76-YAG, Coherent, Palo Alto, CA) with a Regenerative Amplifier (RGA60-10 Continuum, Santa Clara, CA). Pulse duration was 80 ps, beam spot size was 1.1 mm^2 and pulse energy was 18.2 mJ. Time delay in this case was 10.7 ms from the start of the P-pulse. The P-laser parameters were as described above with average power of 12 W and micropulse energy of approximately 40 mJ.

While considering our results for the ablation of dentin, it is possible to argue that enhanced ablation observed with our setup is simply the consequence of tissue water content changes (dehydration) due to the action of the Primer laser. However, in a previous, separate set of experiments [1], we have investigated extensively the ablation characteristics of fresh versus dehydrated tissue during ablation with the same 15 ns XeCl excimer lasers used in these experiments. We have clearly demonstrated that ablation of fresh tissue is actually two to three times as efficient as ablation of dehydrated tissue. If dehydration was the cause of the observed effect, we should have seen the opposite phenomena in wet Vs dry experiments. Additionally, of course, the fact that the same CHAA enhancement is observed in dentin tissue which underwent rigorous chemical dehydration [by consecutive treatment in baths with increased concentration of alcohol, see reference 5], as well as the demonstrated effectiveness of CHAA in the ablation of glass, considerably re-enforces the tendency to discard this hypothesis.

While we now feel rather confident in our ability to show the utility of CHAA and its application potential in a variety of materials, many questions regarding the parameters required to achieve optimal CHAA operation remain to be investigated. For example, the question of optimal depth of light penetration for the primer laser wavelength. Since CHAA is designed to cycle the material at a frequency which is of the order of magnitude of its thermal relaxation time, and τ_r is proportional to δ^2, a deeper optical penetration depth (δ) which on first consideration appears to be a desired property if larger volumes are to be effected, will have to be matched by impulse that effect the same volume. Since deeper δ also imply longer τ_r, deeper δ will also induce small temperature gradient over the optical penetration volume for a longer time and may, thus, not yield the desired differential stress required to compromise material strength.

The question of thermal modulation depth will also have to be resolved by subsequent modeling. Additionally, the optimal time separation between the onset of thermal cycling and the incidence of the impulse is important. Optimal matching of I- and P-laser parameters, and minimal I-laser energy which can still maximize AR will have to be determined as well. Such extensive theoretical and numerical investigations will be rewarding, if the advantages of CHAA over single laser pulse ablation show demonstrated utility.

Finally, we compared our results to other two-lasers ablation enhancement investigations. We note that since the P-laser induces no permanent changes in the tissue it is, therefore, fundamentally different from those described in references 1, 2 and 3. Since the pulse structure of the P laser is entirely different (much longer and repeated micropulse application in our case Vs a single 7 ns pulse in the lasers used by reference 4) the ablation process is quite different. At the same time, we speculate that it is possible that cyclic stressed induced by the first nanosecond laser of reference 4 (and similar to those deliberately induced by our periodic P-laser) may be responsible for the ablation enhancement observed in that experiment.

In conclusion, we have demonstrated a new method for enhancing of ablation rates in organic and inorganic materials. The major advantages of our method are: ablation efficiency by the "impulse" laser is significantly enhanced; "primer" laser is operated under any damage threshold while at the same time lowering pulse energy necessary to achieve damage by the "impulse" laser. Ablation where *both* lasers operate under threshold conditions is also believed possible and will be demonstrated in subsequent reports. Such operation will be useful in enhancing accuracy and precision since ablative effect is achieved only when the two lasers are superposed (this will ensure that damage will occur only in the desired area of overlap). Additionally, reduction in collateral damage and enhanced optical fiber delivery may also be realized during such lower pulse energy and/or subthreshold operating conditions.

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